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MEDICAL ULTRASONIC IMAGING SYSTEMS

FINAL CONTRACTUAL REPORT

W7711-9-7587/001/TOR

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Contract Report: **DRDC Toronto CR 2002-059**

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May 2002

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Executive Summary

Initial contacts confirmed that the DRDC Toronto investigation on 3D adaptive ultrasound beamforming is an important project. It seeks to provide new methods for correcting common artefacts in medical ultrasound images that are caused by variations in the speed of sound in different tissues. The variations can distort images sufficiently to sometimes cause a false finding of twins in o.b. scans. They also confuse the diagnosis of solid versus cystic tumors in other organs, which can falsely indicate a finding of cancer.

The methods proposed in this work use modern signal processing methods to correct the echo signals by using properties of the signals themselves. They have been proven in other applications of pulse-echo imaging such as sonar and radar. The application to the medical ultrasound signals is not trivial because of the much wider fractional bandwidths and higher data rates encountered in this new application.

My role thus, was not only to advise on the medical ultrasound requirements of the system, but also to keep the group informed about current and future applications of ultrasound in medicine. The first task was the evaluation of the proposed design. It was deemed suitable for a "proof of concept" project in that the resolution and other parameters would be capable of producing images that could demonstrate the advantages of the new methods.

During the first visit a number of hardware problems were addressed, and a software program for calculating beam profiles was located for their use. The technical specifications of a three-dimensional scanning system that was under development in the U.S. were also reviewed for comparison. It used very conventional methods for beamforming and was not really relevant to this effort

Effort was also expended in evaluating the proposals for construction of the ultrasonic transducers, a critical element in the operation of the system since they determine the resolution and sensitivity. It was also possible to get some evaluation information for such vendors.

During this time much effort went into methods for correcting the effects of sound speed variations by other research groups. By attending the International Ultrasonics Symposium in Puerto Rico we were able to make contact with these researchers and provide copies of the papers that outlined their methods (called abberation correction in the literature).

Again, they were rather more conventional than those we are developing

During the second trip to Toronto the images produced by realistic simulations were assessed. These images indicated that the effort was making progress toward the goal, but were not yet tested on actual signals from tissue since the subcontractors had not delivered any hardware

During the final visit we stressed that it is critical that actual signals be used for evaluation, and that the work so far indicates that the final goal of producing abberation free three-dimensional image data is within reach. Future plans to reach the goal were discussed in detail.

Finally, this report outlines briefly the conventional ultrasound systems in order to contrast their technology with that the adaptive beamforming innovation being developed at DRDC Toronto by Dr. Stergiopoulos' research group.

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INTRODUCTION

Ultrasound has been called the most widely used imaging modality in medicine. No demonstrable harm has resulted from its diagnostic use, and there are strict regulatory limits on the output. These systems are widely used for primary diagnosis as well as for screening, monitoring and follow-up procedures. The equipment is movable; some even can be carried by hand, and it can have high frame rates. The many features of these systems have been summarized by Greenleaf. [1] The large installed base and experienced personnel almost guarantee that any improvement in image information will benefit patients. Since the frame rates are high enough to follow changes in real time, and the scan head transducers are easily movable, the examinations are very interactive. Many different scan planes can be imaged by moving the small scan heads shown in Figure 1.1. The motions of muscles, organs and even of blood flow can be seen.

The static images in publications, such as those shown here, give a very poor impression of the picture on the display. On the display the heart can be seen beating, muscles and tendons move and the blood flows in pulses. An example of a static image is shown in Figure 1.2. In life the pulsating blood flow in the descending aorta of a fetus in-utero is seen, including a branch to the kidney. A fetus will respond by moving visibly to stimuli so that a neurological examination can be carried out as well. However, experienced personnel are needed to acquire and to save the particular images regarded as diagnostic. This requirement is a current weakness of ultrasound imaging.

These systems offer many opportunities for improvements through signal processing. The speed of operation and extraction of diagnostic information are two promising areas of work. The phase as well as the amplitude of the received signals is available, and most modern systems are fully digital, offering software control of both transmitting and receiving functions (within the limits of the hardware). The received digital signals can be made available for external processing.

Medical ultrasound systems, in common with radar and sonar systems, are active in that interrogating signals are transmitted by a transducer. In our case an electro-acoustic transducer. Small changes in the elastic properties and density of tissue elements reflect the energy back to the receiving transducer. The resolution of the system is set by the dimensions of the emitted and

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received sound beams in both space and time. The overall system response is the product of the separate transmit and receive beams, which need not be identical. The propagation factors must also be included in the response. These factors include tissue attenuation and the diffractive properties of transducer geometry. Medical ultrasound systems have some unique features and limitations that set them apart from radar and sonar systems. At the same time, the differences open new avenues and opportunities for problem solving.

The major differences are:

1. High and frequency-dependent attenuation of tissue leads to a restriction on highest usable frequency, hence depth of imaging and resolution.
2. The tissue of interest can be in the near field of practical transducer apertures so tight focusing is possible..
3. Anatomy sets serious restrictions on access. Bone and lung are more difficult to penetrate than other tissues.
4. Many targets are not discrete objects, so we must map *areas* of echo density and texture.
5. Unresolvable scattering structures give false discrete targets and mask the echoes from small structures through the "speckle" artifact. [2]
6. Only sound speed sets a limit on frame rate, which can be high enough to stop all motions encountered, even in the heart.

Signal processing in this environment can be viewed as having three goals:

1. Faster and better image formation. Avenues include selecting transmitting and receiving elements (beamforming), controlling the transmitted signals, and then processing returns from individual sensor elements in hardware or software.
2. Image enhancement as a second step to increase image quality, and to lessen the need for highly trained operators.
3. Extraction of diagnostic information about tissue, sometimes called tissue characterization, and incorporation of this information into the image. [3, 4]

Current systems perform the first step to make images, and can also do scan conversion to permit display on raster scan monitors. In the second step enhancement processing, such as interpolation of additional image lines, speckle reduction and edge enhancement is done before images are displayed. Details of these processes are highly proprietary.

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The third step includes several processing steps that go beyond basic image formation. One example is to use the Doppler effect and its' relatives to assess blood flow and tissue motion, which are important functional measures. [5] Other efforts to measure elasticity and to derive statistical properties of the scattered energy are showing promise. These added operations require calculations that slow the imaging process with current methods. This added information can be integrated into the images with selection algorithms, and are often shown as an area of color substitution within the basic gray scale, or B-mode, image.

1. SYSTEM FUNDAMENTALS

1.1 RESOLUTION

The lateral resolution is set by the dimensions of the sound beam from the transducer. Consider the beam from an acoustic aperture, shown at the left in Figure 1.3. This beam can be either that transmitted, or the area of sensitivity of the same aperture as a receiver. The wavefronts spread out to the right and form the far field. The lateral resolution is thus poorer at large distances, but improves with a larger aperture. Better resolution closer to the transducer is possible for a smaller aperture. If a small receiving aperture is expanded with time after transmit, this "expanding aperture" system can have a better overall lateral resolution than a system with fixed aperture.

The beamwidth at a range x, along the diverging beam, is given approximately by $\lambda x/D$, which is λ times the instantaneous "F" number of the aperture. In a typical expanding aperture imaging system with a constant receiving F number of 2 or 3, the lateral resolution can be constant, and equal to 2 or 3 λ. This requires that the aperture, Figure, 1.2, be both expanded and focused. At an assumed 4 MHz. frequency, and a sound speed of 1540 meters/sec, the lateral resolution is less than a millimeter. The range resolution can be set to be the same as the lateral resolution. This gives a bit more than 100 resolution elements in either direction. Since both resolution and penetration vary inversely with frequency their ratio is nearly constant, so that the relative resolution is the same in any ultrasound image. The frequency of operation must be chosen to suit the application. Very high absolute resolution is possible at high frequencies, but the penetration is less. For example, the acoustic microscope can approach optical resolution by operating at gigaHertz frequencies. [6]

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In the example above the resolution was improved by focusing at all ranges. However, focusing is only effective in narrowing the beam within the near field of the transducer aperture, since focusing narrows the field to have the same angular width as exists in the far field from the same aperture. In an array transducer system the focusing on receive can be adjusted electronically to track the echo region as the pulse propagates. By this "dynamic focusing" the number of resolution elements can be increased to several hundred if the anatomy allows a sufficiently large aperture. In some patients the variable sound speed in tissues, mainly inclusions of fat, can upset the precise timing needed to form very narrow beams. This results in beam distortions that can misplace structures in the image, or can fill normally echo-free cysts with low level echoes. This makes the important cystic vs. solid tumor decision more difficult. The correction of these "phase aberrations" is an active area of research.

1.2 SCANNING AND TRANSDUCERS

The transducer is a critical component of the imaging system. [7] Its' size, resonant frequency and bandwidth determine the resolution and penetration. Many types have been developed to provide desired images in a wide variety of medical applications. These include both single element types needed for high frequencies, and the multi-element arrays shown in Figure 1.1.

The basic scanning array, sometimes called a "1-D" array, is shown in Figure1.4. The array is formed by slicing the radiating structure into independent elements with widths of the order of the center frequency wavelength in tissue or water. Electrical connections are made to each element. The transmitting and receiving beams of ultrasound are formed and scanned by the associated electronic circuits. [8]

Because of the Fourier transform relationship between the radiating aperture size and the beamwidth, the resolution is better in the scanning plane, Figure 1.4, than in the transverse, or cross-plane, direction. Efforts are being made to reduce this effect, which can produce confusing echoes from structures that are out of the image plane. The poorer transverse resolution also results in the reception of more clutter signal energy, which makes the speckle artifact worse.

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A current development is the use of elevation focusing of arrays to narrow the thickness of the imaged slice of tissue. A simple weak focusing lens of a rubber with a sound speed less than that of tissue has been used for this purpose for some time, since it also provides a curved front surface that is comfortable for the patient. With weak focusing the focal region in the elevation plane is cigar-shaped, since then the far field contracts toward the axis at long ranges.

Better focusing in elevation can result from dividing each array element into several sections, as shown in Figure 1.5. Separate connections to these parts of the element allow expanding the aperture using switches (called a 1.25 D array), or allow focusing in elevation using delays (called a 1.5 D array). A more finely divided array element could be programmed to scan over a limited range in elevation. This is called a 1.75 D array. [8] The ultimate solution to the coarse focusing in elevation is the fully two-dimensional array, with equal numbers of elements in azimuth and in elevation. In this case the beam can be directed freely over a three dimensional (3-D) region of tissue, to collect a three dimensional data set.

The goal of three dimensional data collection is being approached in three ways. These are: by post-processing of individual scan images; by sweeping a 2 D scan plane in the transverse direction by manual or mechanical means and storing the data in real time; or by using a true 2 D array. One true 3-D system of this last type exists and more may be expected. [9] They are most useful for imaging complicated 3-D structures such as the heart and the fetus, in which the plane "slice" that is imaged by any one scan cuts through complicated structures and is difficult to interpret. In these cases the goal is to display a slice at an orientation that shows the desired structures. A suitably diagnostic scan plane often can be found after searching the 3-D data set. The examination takes only a few minutes for data collection, and removes the need for most repeat scans.

The current array transducers use several scan formats as shown in Figure 1.6, and the system may be required to accommodate more than one. (a) is the rectilinear scan as used in small parts and some abdominal imaging. The beam only need be translated, not phased, to form this format. The dotted lines show regions where the transmit beam can be focused. Since these structures move only slowly repeat scans can be made with different fixed

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positions for the transmit focus, and the image then is assembled from the focal regions of the scans.
(b) is the scan from a curvilinear array. This scan increases the field of view over that of the rectilinear scan, and also only requires simple translation of the beam. Expanding aperture and dynamic focus operation is still possible but the number of elements that can contribute to any one scan line is limited by the curvature.
(c) shows the sector scan that is used with a phased array for imaging the heart, or any structure where the imaging "window" is limited by anatomy. Here, mechanical rotation or time delays are used to scan in angle. Some very strongly curved linear arrays can produce scans in this format.
(d) is the fully rotational scan used for intravascular imaging, either by mechanical rotation of a single element or by a circular phased array.

2. TISSUE PROPERTIES' INFLUENCE ON SYSTEM DESIGN

2.1 ATTENUATION

The tissue environment is one of relatively high and frequency-dependent attenuation, which has a profound effect on system design. These properties have been tabulated in some detail. [10] Average soft tissue has a one-way loss of approximately 1 dB per cm. for each MHz. of frequency, which ranges from about 1 to 40 MHz, depending on the application.. Lower values of attenuation that are sometimes found in the literature were adopted to make conservative estimates of safety from heating and cavitation, but do not govern the imaging properties. The presence of attenuation means that the frequency cannot be selected arbitrarily, since both the penetration and resolution depend on the choice.

At 4 MHz. the loss over a 10 cm. range could be over 80 dB, plus enough reserve sensitivity to handle 30- 60 dB reflection loss, or maybe 140 dB. A receiver gain function that increases with time after transmitting is needed to avoid overload in the receiver. This is not just a problem with analog systems.

In a digital system the problem is to stay in range of the A/D converter. A dynamic range of over !40 dB needs over 23 bits. Modern systems need digitization rates to 20 MHz or more, and this hardware combination is now impractical. Therefore a controlled gain function (AGC) is still needed in the

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receiver to compensate for attenuation. Currently the rate is set by the operator. The amplifiers are followed by anti-aliasing filters and 8 to 12 bit flash or pipeline A/D converters. The converters must be located in the channel connected to each active receiving element. Since one bit is lost with a.c. signals and another lost to noise, the total dynamic range from the remaining bits could still be insufficient to handle the expected 60 dB dynamic range from echo variations alone. Since the outputs of many individual A/D converters are added together in the usual delay and sum beamformer, a longer word length results which can help to meet the requirement.

2.2 SPEED

The frame rate is limited only by the speed of sound, c. Fortunately soft tissues have nearly the same sound speed. Fat alone deviates from the average by about 10%. Frequency dispersion is similarly low. At an average speed of 1540 meters/sec. the ranging time, 2x/c, is 13 μ sec./ cm. The ranging time for radar is 12.2 μ sec / nautical mile, so, considering the different distance scales, the timing electronics and the processor speeds are similar. Our pulse repetition rate (p.r.f.) is a few kHz. for abdominal imaging. Imaging time is proportional to the number of actual lines in the image. Sixty frames per second is about the theoretical maximum for a 128 line system with 10 cm. range. A lower rate is needed for retrace allowance, but single scan planes can be imaged rapidly enough to stop tissue motion.

For 3-D imaging another strategy must be used to keep the frame rate high enough for the heart. The needed multiple channels have been formed by using receiving channels that synthesize multiple receive beams for each pulse transmission. One 3-D system has a rate of 22 full 3-D data sets per second and uses 16 simultaneous receiving channels. [9]

2.3 STRUCTURE

Tissue structure can be viewed as being heterogeneous on any size scale. This leads to the production of echoes throughout the volumes we wish to image. The dimensions of the resolution cells can be viewed as forming a sample volume from which both desired and undesired, or clutter, signals are received. Since the clutter elements seem to be randomly distributed their echoes add with random phases and lead to the formation of speckle, just as in radar and

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sonar. The amplitude peaks of the video speckle signal can mask small structures and break up organ outlines.

The dimensions of the speckles are the size of the resolution elements in the image plane, so are useful in judging the resolution of a system. Expanding aperture systems strive for a constant speckle size, but the speckles do get larger at the maximum range after the full aperture is reached. Another use for speckle is to calculate the speed of an echo from a moving structure independently of its' direction. The differences in speckle location from image to image are mapped as a velocity field. Correlation of one image with another can be done on either the r.f. or the video signal.

Both frequency and spatial compounding have been used to reduce the speckle artifact with modest success. The full 3-D systems have an advantage in being able to produce a very small sample volume, since the out-of-plane resolution can be made the same as that in the image plane. This results in less speckle energy. Many statistical methods of reducing speckle have been investigated without much success. They usually require signals from regions that are larger than the smallest possible resolution elements. The processing tends to blur edges and structures.

3. IMAGING SYSTEMS

3.1 SINGLE CHANNEL

The basic principles are the same for single and multiple channel systems, and will be described first. Originally, mechanical scanning was used in single channel analog systems. A typical functional block diagram is shown in Figure 3.1. These systems are still used at the highest frequencies where array fabrication is not possible because the saw kerfs used to subdivide array elements remove too much material. These single transducer systems are used in the acoustic microscope, scanning inside small blood vessels and for scanning skin and the eye.

The transmitter emits short, wideband pulses of only a few cycles. With modern, high bandwidth, transducer materials the resolution goal of a few wavelengths is easily met. The repetition rate is a few kiloHertz, chosen to be low enough to avoid "second time around" echoes. A duplexer is used to protect the receiver so that it can recover full gain rapidly after the overload

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from the transmitter. Passive circuits are used here, since active switches have too much spurious output from switching transients.

The r.f. portion of the receiver has two main functions:

1. To amplify the weak echoes with a gain that increases with time (AGC) to compensate for attenuation.
2. To filter to select the signal bandwidth, to prevent aliasing and to reject noise.

Post-detection digital techniques are now common in high-frequency systems and are also used in low-end equipment. The A/D converters are located just following the detector shown in Figure 3.1, and operate at the lower frequencies of the video signals. The introduction of digital techniques allowed digital image storage, which also provided scan converter operation for TV raster display. The associated software control allows some signal processing to be done digitally. Examples are interpolation of image lines, sweep integration, contrast adjustment and edge enhancement. The digital output from these systems is only the detected and filtered video display signal.

In the common scanner designs meant for observing blood flow, the Doppler shift is extracted at a single location in the image, and is interpreted as the flow velocity component in the direction of the sound beam. The spectral display shows the direction of flow and can indicate the presence of turbulence as well. The receiver incorporates demodulation to baseband in in-phase (I) and quadrature (Q) video channels, see Figure 3.2. Sample and hold circuits select signal amplitudes at one range at the pulse repetition rate, and filter to remove strong signals that result from motion of the vessel walls, and to correct the frequency response. This complex video signal is fed to an FFT analyzer to find the Doppler shift at that range. The resulting Doppler spectrum is usually presented as a function of time along with an image showing the location where the Doppler shifted signal originates, see Figure 3.3. [5]

The phase is sampled at the p.r.f., which limits the maximum unambiguous frequency shift in each direction to one-half the p.r.f. Since the systems are directional this is a total frequency span equal to the p.r.f. Since the zero frequency can be offset by a small change in the reference frequency the upper limit of Doppler shift measurable without aliasing equals the p.r.f.

Since there may also be strong signals from the vessel walls a high-pass "wall filter" is needed before the A/D to remove them. This filter cutoff

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frequency may be selectable so that the operator can reduce its' effect when observing low flows. In some designs a low cutoff frequency may remain when the control is set to "zero".

3.2 MULTI-CHANNEL SYSTEMS, ARRAYS

Figure 3.4 shows the basic electronic system used with modern aperture arrays. The active elements of the array used for transmitting and receiving are selected by the multiplexer. Switching is done during the retrace time to avoid transients. The number of receiving and transmitting channels may be the same or less than the number of transducer elements. The transducer is connected to the transmitter and the receiver through their respective beamformers, see Figures 3.5 and 3.6. The beamformers introduce any time delays needed for focusing and adjust the amplitudes (apodization) to reduce sidelobes in the radiation pattern. Grating lobes, which result from a too wide array element spacing are eliminated by using a fine pitch in the array, or reduced by differing transmitting and receiving beam patterns in which the grating lobes do not overlap.

In the basic linear array the multiplexer determines which elements will transmit and receive, and it changes which elements are active to scan. If the array has a flat surface the scan format in Figure 1.6 (a) results. In a curvilinear array the curvature of the front face provides angular sweeping of the beams, and the scan formats of Figure 1.6 (b) or (c) result.

The beamformers for so-called phased arrays do beam steering and focusing by introducing suitable time delays.[8] Phasing is a narrow-band method that is only useful for fine, or vernier, delays in this system.

The array receiver beamformer is shown in more detail in Figure 3.6. It can adjust the focusing delays with time after transmission (dynamic focusing). This latter technique tracks the received focal spot as it moves through the tissue at the speed of sound to remove any depth of field problem with the focus, and attempts to maintain a constant "F" number for the system.

Each channel of the beamformer, Figure 3.6, contains all the needed receiver elements except for the detector and video processors. Many AGC amplifiers are needed to keep the A/D converters within range. The amplifiers also contain duplexers and anti-aliasing filters. The A/D converters must handle

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twice the r.f. frequency. The processing in the array systems must provide some filtering to remove low frequencies introduced by the differing d.c offsets in the A/D converters. This may be done either in hardware or software. Again, quadrature detection, as shown in Figure 3.2 but operating on all of the received signals, can use A/D converters operating at just over twice the signal bandwidth rather than at the highest r.f. frequency. The problem remains that it is quite expensive to provide more than 64 or 128 channels. So-called "synthetic aperture" systems avoid this problem.

The synthetic aperture approach was introduced in commercial systems to produce systems with the performance of many channels without actually having to build them. Here, this term means time-sharing of a number of transmitter and receiver channels to synthesize the electronic system. The full physical aperture is already provided by the transducer. More advanced synthetic aperture processing is being developed, and is a promising field, but better hardware is needed. [11]

The basic block diagram illustrating a fully digital system suitable for synthetic aperture operation is shown in Figure 3.7. Two important new functions have been added. First, the data collected by firing the smaller number of transducer elements and collecting the returns from the active receive channels are stored. The signals can be at the original r.f. frequency, or in I and Q channels if down-converted to baseband by quadrature detection. Thus a large high-speed memory is needed. Then all of the original functions needed for image formation are done in a processor, using software.

The transmit beamformer structure is not changed, but the receiver is a bit more simple, see Figure 3.8. The output of the A/D converters now goes directly to the data memory. Filtering for the d.c. offsets is still needed. The functions of the processor, Figure 3.8, can include selecting the data from a selected number of elements for expanding the aperture, performing apodization and adjusting the timing of the data lines for scanning and focusing. Since the r.f. data are available, Doppler processing in software can be performed with this same hardware.

New types of processing are now possible in this system. Theoretically, a new focused transmitter firing can be done for each image pixel, and the echo data collected from each focused transducer element, so that focusing of the full array can be done on transmit as well as on receive. This truly confocal

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system has been called the "gold standard" beamformer. For an "N" element array only $N^2/2$ firings are needed since reciprocity holds. [8] In applying any form of signal processing on data from an ultrasound system it should be recognized that significant processing may be inherent in the many possible types of operation. Collaboration with the manufacturer generally is needed to be able to determine what has been done, and how it may affect data acquisition.

3.3 DOPPLER AND ADVANCED SYSTEMS

When the I and Q signals from quadrature detection are present Doppler shift extraction can be done for all pixels in the image since phase information is preserved. With suitable algorithms the rate of change of phase can be calculated at any range. From this information the mean Doppler frequency and other parameters, such as its variance, can be calculated and displayed as velocity and turbulence, respectively. The mean Doppler frequency is usually found from the phase of the complex autocorrelation function as

$$\varpi = \frac{1}{\Delta T}\tan^{-1}\frac{\sum_{i=1}^{n} I(i)Q(i-1) - Q(i)I(i-1)}{\sum_{i=1}^{n} I(i)I(i-1) + Q(i)Q(i-1)} \qquad (1)$$

where the index, *i*, refers to *n* successive image lines spaced ΔT apart. [12]

Since the scan line data also includes the much stronger tissue echoes it is necessary to precede the Doppler processor by a stationary echo canceller, similar to those used in radar. This serves the same function as the wall filter in single channel systems.

Digital control of the display is also needed to allow selection of the region where the Doppler output, or color flow, signals are seen in shades of color to distinguish them from the gray tissue image. This region is usually smaller than the full image to allow time for the additional processing, which requires several transmit bursts. The selection of which data set to display is done on the basis of amplitude, since the echoes from red cells are quite weak.

The scan image in Figure 3.3 (a) is an example of this display. The B mode image of the uterus without the Doppler component is at 3.3 (b). The

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added information clearly shows the presence of high speed turbulent flow that is found at arterio-venous fistulas.

An alternate method of Doppler display is called power Doppler. [5] Only the magnitude or presence of the Doppler signal is displayed, not its frequency, and the signal is sampled only at its' peak in the cardiac cycle. The image requires a few heartbeats to develop fully, and is useful for imaging regions of low speed flow in small vessels and for revealing non-echogenic plaques on vessel walls. Figure 3.9 shows how power Doppler can delineate the vessels in a liver.

Algorithms for displaying low flows take advantage of the high scanning speed of electronic scanning. Following radar practice, the early mechanical scanners collected the Doppler data from a small number of firings in each look direction, before moving to the next, slowing the scan rate. This method collects enough data for a number of images, equal to the number of transmitter firings, which the electronically scanned array can produce. With the faster imaging rate the pixel sampling rate decreases to p.r.f./n, where n is the number of look directions, and the sample window is wider.

A third type of Doppler processing is proving to be useful. In this "tissue Doppler" the wall filter or stationary echo canceller is removed, and the display then shows the echoes from moving tissues in color. The motions of the myocardium can be seen better in this display. The display is also useful in assessing plaques and blockages in arteries.

Imaging using the nonlinear properties of tissue and contrast agents has recently been introduced. Three types of operation are available: harmonic, subharmonic and pulse inversion imaging. The first uses higher harmonics of the transmitted frequency. The transmitter operates at a given frequency and the receiver is tuned to the second harmonic. Nonlinear propagation of finite amplitude ultrasound produces the harmonics which increase in strength with depth. The resulting images are superior to those obtained by imaging at the fundamental, for reasons that are not fully understood. Since a lower transmit frequency can be used, image distortions caused by sound speed variations are lessened, while the better resolution for the harmonic components improves the image. The depth of penetration is a bit less with current embodiments, but the image quality and diagnostic utility has impressed the clinicians.

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The second type of nonlinear imaging uses the reflections at one-half of the transmitted frequency; the subharmonic. Since tissues produce a very low level of subharmonic energy the contrast echoes are more easily seen in a tissue background.

A particularly effective form of nonlinear imaging can also be done if the hardware permits. In this "pulse inversion" imaging the polarity of the transmitted pulse is changed between two transmit firings. [13] The received pulse trains from sequential firings are added in the receiver, so that the fundamental frequencies in the echoes are cancelled and only the non-linear components are used for imaging. This method can include the subharmonics.

These nonlinear methods offer great promise for use with contrast agents. Ultrasound contrast agents contain small gas bubbles, which are non-linear scatterers and produce strong reflections as well. The echoes from the fluid in the very smallest vessels are often undetectable without the use of contrast. Harmonic methods are capable of enhancing the contrast echo since the bubbles are more nonlinear than the tissues.

Coded pulse transmissions and suitable receive processing is also being done in at least one commercial system. This allows the sensitivity to be increased without increasing the peak power and hence the possibility of cavitation. [14] With this system deeper penetration is possible for a given frequency. The echoes from red cells can be seen on the image. A selection algorithm is needed to control the regions of the image where echoes from blood are shown, just as in Doppler or color flow imaging.

One disadvantage of ultrasonic imaging is the small region that is imaged. It is difficult to include "landmarks" in the image which allow persons other that the operator to interpret the image. It is also difficult to perform registration with images from other imaging modalities. It may even be difficult to compare ultrasonic images from adjacent regions for diagnostic purposes. It has been possible to use image data itself to allow adjacent scans to be connected to form a larger image. [15] The principle is shown in Figure 3.10. An algorithm based on "fuzzy logic" is used to connect the separate images without the need for any mechanical constraint of the scan head. Another example of this SieScape™ processing is shown in Figure 3.11, which shows how a necrotic abdominal wall mass can be shown not to involve the liver.

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4. CONCLUSION

For our use in doing signal processing, we now have systems with stored digitized data that can be made accessible to us. Software control of processing is built into the machines and can be used for many purposes. Many new schemes are now possible, including creating multiple receive beams for each conventional firing to increase the frame rate, and special processing to correct for phase aberrations and speckle degradation of images. In the most modern systems some control of the transmit waveform on each element is possible, although care must be used to stay within safety guidelines. The received signals from each element are sent to a digital memory since many image forming and enhancing operations are now done in software. The manufacturers can make these data available for processing, but close cooperation is necessary.

We now have the opportunity to apply even more advanced methods of signal processing to improve the utility of these systems. This includes increasing operating speed, lessening dependence on trained operators, improving image quality and increasing diagnostic accuracy.

References

1 Greenleaf, J. F., Acoustical medical imaging instrumentation, in *Encyclopedia of Acoustics,* Crocker, M. J., Ed., John Wiley and Sons, New York, 1997, vol. 4, chap. 144.

2. Wagner, R. F., Smith, S. W., Sandrik, J. M., and Lopez, H. Statistics of speckle in ultrasound B scans, *IEEE Trans. Sonics Ultrason., 30, 156, 1983.*
3. Greenleaf, J. F., *Tissue characterization with ultrasound,* vols I and II, CRC Press, Boca Raton, FL., 1986.
4. Shung, K. K., and Thieme, G. A., *Ultrasonic scattering in biological tissues,* CRC Press, Boca Raton, FL., 1993.
5. Evans, D. H., *Doppler ultrasound: physics, instrumentation and clinical applications,* John Wiley and Sons, New York, 1989.
6. Lemons, R. F., and Quate, C. F., Acoustic microscopy, Chapter 1 in *Physical acoustics; principles and methods,* vol xiv, Mason, W. P., and Thurston, R. N., eds.New York, Academic Press inc., 1979.

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7. Reid, J. M. and Lewin, P. A., *Ultrasonic transducers, Imaging*, Wiley Encyclopedia of Electrical and Electronics Engineering,. Webster, J., Ed., vol 22 p. 664., 1999.
8. Thomenius, K., Evolution of ultrasound beamformers, *Proc. 1996 IEEE Ult. Symp.*, San Antonio, TX,1615,1996.
9. Light, E. D., Davidsen, R. E., Fiering, J. O., Hruschka, T. A., and.Smith, S. W., Progress in 2-D arrays for real time volumetric imaging, *Ultrasonic Imaging*, 20, 235,1998.
10. Goss, S. A., Frizzell, L. A., and Dunn, F., Ultrasonic Absorption and attenuation in mammalian tissues, *Ultrasound in Medicine and Biology*, 5, 181, 1979.
11. Hazard, C. R., and Lockwood, G. R., Theoretical assessment of a synthetic aperture beamformer for real-time 3-D imaging, *IEEE Trans. on Ultrasonic, Ferroelectrics and Frequency control*, 46, 972, 1999.
12. Kasai, C., Namekawa, K., Koyano, A. and Omoto, R., Real-time, two-dimensional blood flow imaging using an autocorrelation technique. *IEEE Trans. Sonics and Ultrasonics* SU-32, 458, 1985.
13. Simpson, D. H., Chin, C. T., and Burns, P. N., Pulse inversion Doppler: a new method for detecting nonlinear echoes from microbubble contrast agents, *IEEE Trans. on Ultrasonic, Ferroelectrics and Frequency control*, 46, 372, 1999.
14. Haider, B., Lewin, P. A., and Thomenius, K., Pulse elongation and deconvolution filtering for medical ultrasound imaging, *IEEE Trans. on Ultrasonic, Ferroelectrics and Frequency control*, 45, 98, 1998.
15. Weng L., Tirumalai A. P., Lowery C. M., Nock L. F., Gustafson D. E., Von Behren P. L., Kim J. H. US extended-field-of-view imaging technology. *Radiology* , 203(3), 877, 1997.

Figure 1.1 Representative ultrasonic scan heads (array transducers). Clockwise from top: linear array, curvilinear abdominal array, high frequency linear, another curved abdominal, curved intravaginal, two phased arrays, highly curved array, side-looking linear array for

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intraoperative use, and microcurved intravaginal array. Courtesy of ATL Ultrasound.

Figure 1.2 Ultrasonic image of a fetus in-utero. The central portion is a color flow image of the fetal aorta. In the actual examination fetal motion and the pulsating blood flow are seen. Courtesy of ATL Ultrasound.

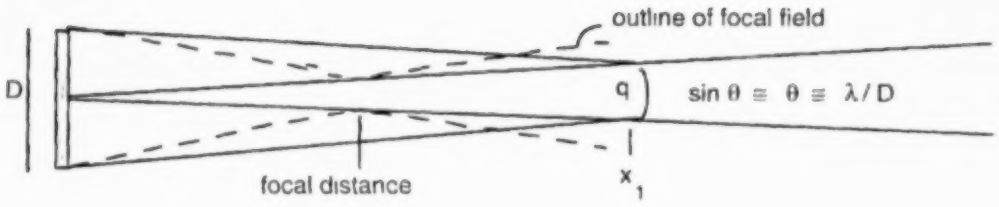

Figure 1.3 Schematic cross section of the approximate extent of the transmitted field from (or the receiving sensitivity of) a transducer aperture of dimension D, at the left. The far field spreads out with the divergence angle, θ, at long distances from the aperture. The minimum field size occurs at x_1, the transition between the near and far fields. An assumed focal field is shown dotted. Focusing can only narrow the field to the same divergence angle, θ.

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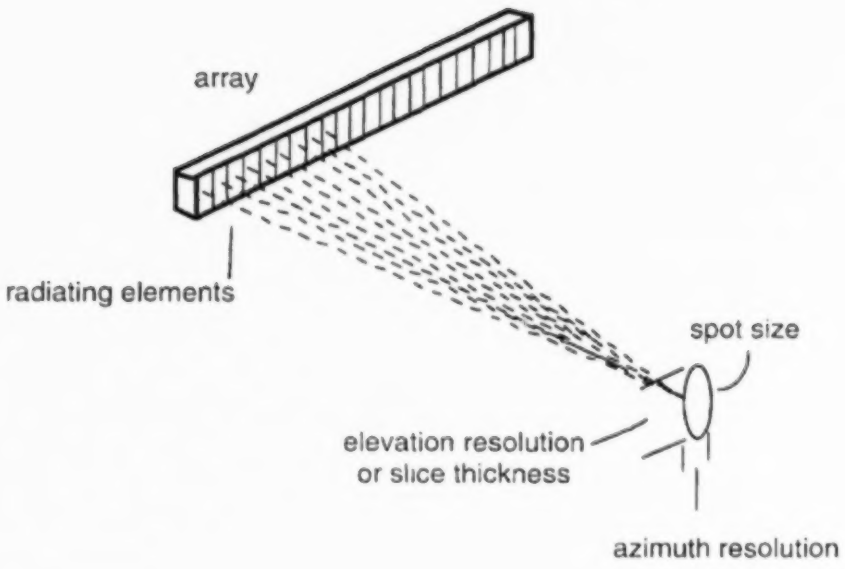

Figure 1.4 Drawing of a multi-element array transducer, showing the path of sound beams from individual elements forming the scanning spot. Electronic control of time delays for excitation and reception form the scanned and focused spot as shown.

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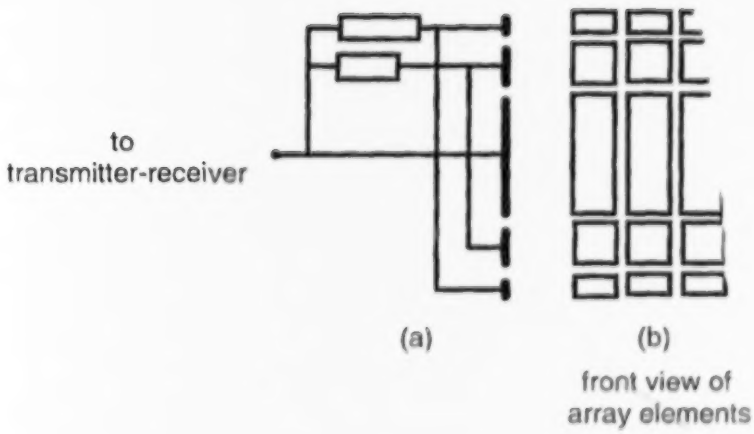

Figure 1.5 Elevation focusing with subdivided array elements. Control of focus is accomplished by using switches or delay elements within the blocks. From Transducers; Imaging, J. M. Reid and P.A. Lewin, in Wiley Encyclopedia of Electrical and Electronics Engineering, Reprinted by permission of John Wiley and Sons, ©1999.

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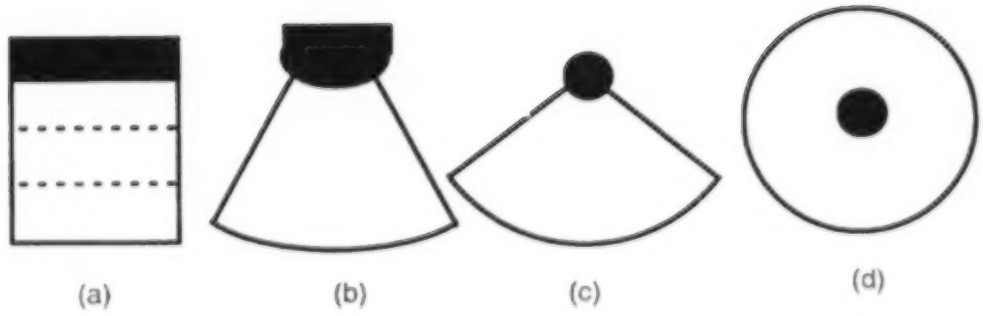

Figure 1.6 Image formats. (a) rectilinear scan, (b) scan from a curvilinear array, (c) sector scan, (d) rotational scan. See text. From Transducers; Imaging, J. M. Reid and P.A. Lewin, in Wiley Encyclopedia of Electrical and Electronics Engineering, Reprinted by permission of John Wiley and Sons, ©1999.

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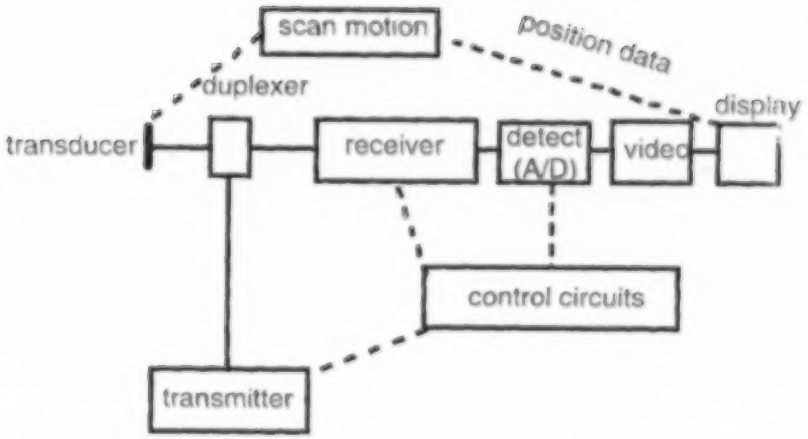

Figure 3.1 Single channel system block diagram. This mechanical scanning system is used with high-frequencies where array fabrication is impractical.

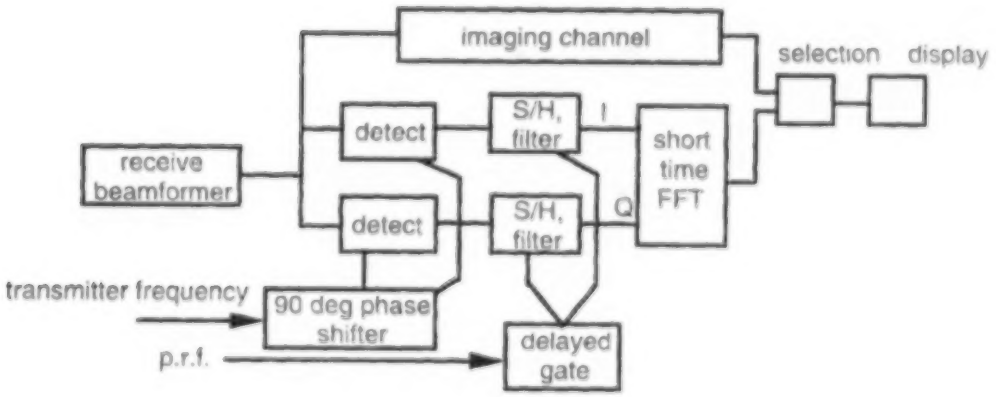

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Figure 3.2 Block diagram of a spectral Doppler processor, using quadrature detection to derive in-phase (I) and quadrature (Q) components for FFT processing. See figure 3.3 for an example of the display.

Figure 3.3 Scans of a uterine growth. (a) Doppler scans of a suspicious region at the head of the uterus, (b) original B mode scan. The color flow sector in (a) shows very high speed flow, by reference to the color scale at the right. The Doppler spectrum from a sample volume placed within a high flow region of the image is shown at the bottom as a function of time. The frequencies, converted to flow velocities by reference to the direction of the line segment in the image, show a uniform distribution of flow speed in both directions from near zero to the maximum. This indicates simultaneous flow in both direction, which is characteristic of tangled vascular beds or turbulence. Courtesy of ATL Ultrasound.

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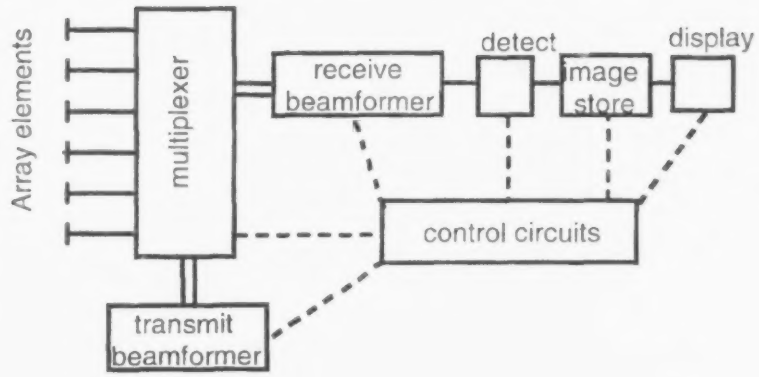

Figure 3.4 Array system block diagram. A multiplexer is needed to connect the transmit and receiving channels to the array elements.

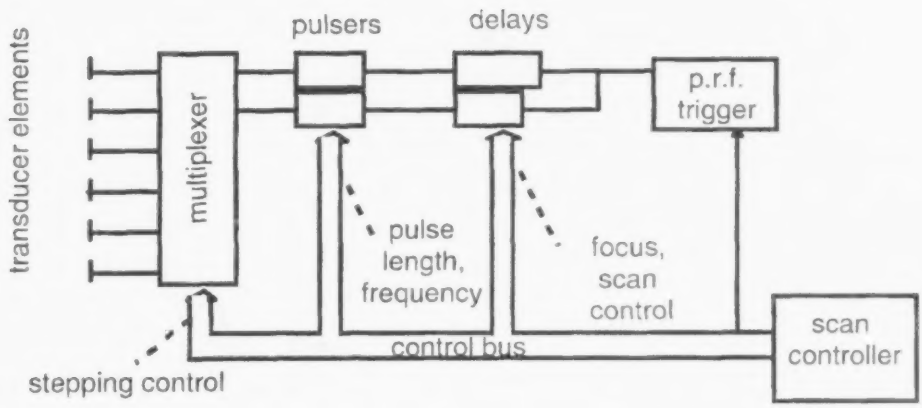

Figure 3.5 Detail of the transmit beamformer of figure 3.4. Note that simple trigger delays are used for focus and scanning.

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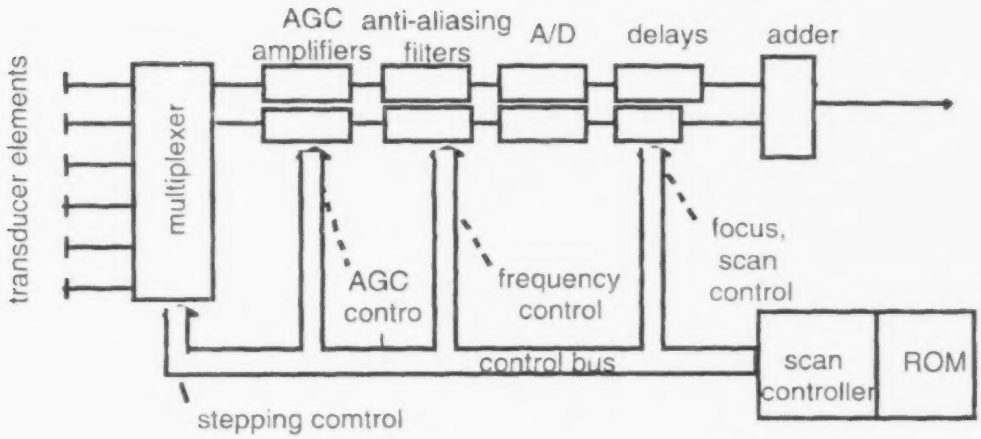

Figure 3.6 Detail of the receive beamformer of figure 3.4. Here wideband or digital delays are used.

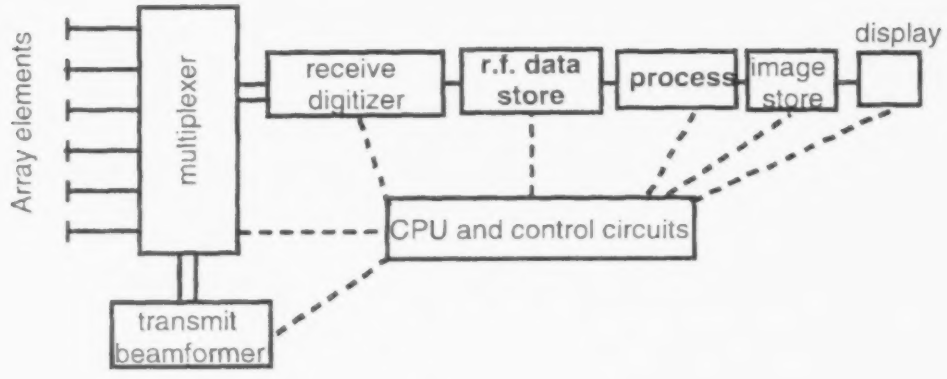

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Figure 3.7 Block diagram of a fully digital system, suitable for synthetic aperture operation.

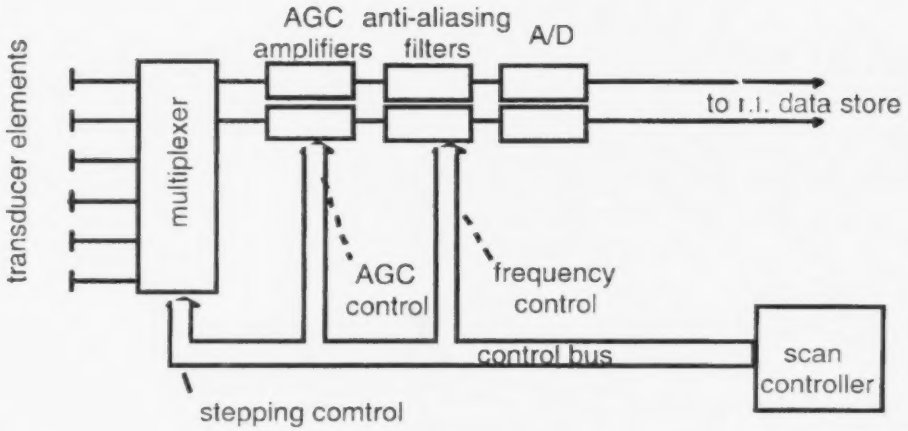

Figure 3.8 Detail of receive digitizer for system in figure 3.7. All beamforming and processing steps are done in software.

Figure 3.9 Power Doppler image of flow in the liver, showing flow in small vessels. Courtesy of ATL Ultrasound.

Figure 3.10 Schematic showing the SieScape™ process of assembling a large field of view image from individual images taken by sliding the transducer along the leg. The image of the saphenous vein at the bottom shows an open flow channel. This scan was not confined to be in a plane, but followed the path of the vein. Courtesy of Siemens Medical Systems, Inc., Ultrasound group.

Figure 3.11 This clinical SieScape™ image shows the relationship of a necrotic metastatic abdominal wall mass to the liver, all on one image. Courtesy of Siemens Medical Systems, Inc., Ultrasound group.

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DOCUMENT CONTROL DATA SHEET

1a PERFORMING AGENCY	2 SECURITY CLASSIFICATION
John M. Reid, 16711 - 254th Avenue S E , Issaquah, Washington 98027	UNCLASSIFIED Unlimited distribution -
1b PUBLISHING AGENCY DRDC Toronto	

3 TITLE

(U) Medical ultrasonic imaging systems

4. AUTHORS

John M. Reid

5 DATE OF PUBLICATION	6 NO OF PAGES
May 1 , 2002	23

7 DESCRIPTIVE NOTES

8 SPONSORING/MONITORING/CONTRACTING/TASKING AGENCY
Sponsoring Agency.
Monitoring Agency
Contracting Agency DRDC Toronto
Tasking Agency

9 ORIGINATORS DOCUMENT NO	10 CONTRACT GRANT AND/OR PROJECT NO	11 OTHER DOCUMENT NOS.
Contract Report CR2002-059	6ca	

12 DOCUMENT RELEASABILITY

Unlimited distribution

13 DOCUMENT ANNOUNCEMENT

Unlimited announcement

14 ABSTRACT

(U) Initial contacts confirmed that the DRDC Toronto investigation on 3D adaptive ultrasound beamforming is an important project. It seeks to provide new methods for correcting common artefacts in medical ultrasound images that are caused by variations in the speed of sound in different tissues. The variations can distort images sufficiently to sometimes cause a false finding of twins in o.b. scans. They also confuse the diagnosis of solid versus cystic tumors in other organs, which can falsely indicate a finding of cancer.

The methods proposed in this work use modern signal processing methods to correct the echo signals by using properties of the signals themselves. They have been proven in other applications of pulse-echo imaging such as sonar and radar. The application to the medical ultrasound signals is not trivial because of the much wider fractional bandwidths and higher data rates encountered in this new application.

My role thus, was not only to advise on the medical ultrasound requirements of the system, but also to keep the group informed about current and future applications of ultrasound in medicine. The first task was the evaluation of the proposed design. It was deemed suitable for a "proof of concept" project in that the resolution and other parameters would be capable of producing images that could demonstrate the advantages of the new methods.

During the first visit a number of hardware problems were addressed, and a software program for calculating beam profiles was located for their use. The technical specifications of a three-dimensional scanning system that was under development in the U.S. were also reviewed for comparison. It used very conventional methods for beamforming and was not really relevant to this effort.

Effort was also expended in evaluating the proposals for construction of the ultrasonic transducers, a critical element in the operation of the system since they determine the resolution and sensitivity. It was also possible to get some evaluation information for such vendors.

During this time much effort went into methods for correcting the effects of sound speed variations by other research groups. By attending the International Ultrasonics Symposium in Puerto Rico we were able to make contact with these researchers and provide copies of the papers that outlined their methods (called abberation correction in the literature).

Again, they were rather more conventional than those we are developing.

During the second trip to Toronto the images produced by realistic simulations were assessed. These images indicated that the effort was making progress toward the goal, but were not yet tested on actual signals from tissue since the subcontractors had not delivered any hardware.

During the final visit we stressed that it is critical that actual signals be used for evaluation, and that the work so far indicates that the final goal of producing abberation free three-dimensional image data is within reach. Future plans to reach the goal were discussed in detail.

Finally, this report outlines briefly the conventional ultrasound systems in order to contrast their technology with that the adaptive beamforming innovation being developed at DRDC Toronto by Dr. Stergiopoulos' research group.

15 KEYWORDS, DESCRIPTORS or IDENTIFIERS

(U) ultrasound systems; A-scan; B-scan; C-scan; ultrasound probes; transducer arrays; abberation effects

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CA021105